ASYMMETRIC ULTRA-SHORT GRADIENT COIL FOR MAGNETIC RESONANCE IMAGING SYSTEM

DESCRIPTION

The following relates to the magnetic resonance arts. It finds particular application in magnetic resonance imaging, and will be described with particular reference thereto. However, it also finds application in magnetic resonance spectroscopy and other magnetic resonance techniques.

5   Magnetic resonance imaging scanners with magnet bores that are short in the axial or z-direction reduce patient claustrophobia and can provide improved access to the patient for interventional procedures. A short bore magnet may, for example, have bore length of 1.0 meters or less.

In short bore magnets, the bore diameter is not scaled downward with the

10   shortening of the bore. A minimum bore diameter is imposed by the cross-section of the imaging subject. Moreover, a reduced bore diameter contributes to patient claustrophobia. Thus, the short bore magnet preferably has a small length-to-diameter ratio. For example, a bore length that is shorter than the bore diameter can be advantageous in a short bore magnet.

15   If the length of a gradient coil assembly is made smaller, especially if the length-to-diameter ratio is 1.1 or less, the efficiency of the transverse gradient coils becomes worse. This is because the return conductors of the saddle coils from which these coils are made up have to be located closer to the working volume of the coil; this leads to a concentration of magnetic field near the conductors of the gradient coil and hence to an increase in the

20   magnetic stored energy of the coil. The peak power to be delivered to the coil in order to generate a certain gradient field amplitude in a certain time is directly proportional to the coil's stored energy. Thus, in order to limit the size and the cost of the gradient amplifier it is important to minimize the magnetic stored energy of a coil. For the gradient coil generating the gradient field in the z-direction the reduction of efficiency on reducing the

25   length is far less severe than for the transverse coils.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect, a magnetic field gradient coil is disclosed for a short-bore magnetic resonance imaging scanner. Upper and lower magnetic field gradient coil windings define a subject-receiving bore and generate transverse magnetic field gradients imposed on an associated static magnetic field generally oriented in a longitudinal direction through the bore. The upper gradient coil winding have an arcuate curvature (C_{upper}) transverse to the longitudinal direction and a longitudinal length (L_{upper}) in the longitudinal direction that is smaller than a longitudinal length (L_{lower}) of the lower gradient coil winding.

According to another aspect, a magnetic resonance imaging scanner is disclosed. A housing has (i) an imaging volume imaged by the scanner and (ii) an imaging subject support section disposed below the imaging volume. The imaging subject support section extends beyond a length of a magnet bore. A radio frequency coil is arranged to inject a radio frequency signal into the scanner bore. A magnetic field gradient coil includes a lower section disposed in the imaging subject support section of the housing and an upper section that together with the lower section define a coil bore containing the imaging volume. The upper section has an arcuate curvature and coil windings spanning a first length. The lower section has coil windings spanning a second length greater than the first length.

According to yet another aspect, a method of magnetic resonance imaging is provided. A main magnetic field is generated through a subject receiving bore. Magnetic field gradients are generated across the bore with a combination of (i) upper gradient coils that have a first longitudinal length (L_{upper}) shorter than a diameter (D_{bore}) of the bore and (ii) lower gradient coils that have a second longitudinal length (L_{lower}) longer than the diameter of the bore. The first longitudinal length (L_{upper}) is generally shorter than a length of the bore and the second longitudinal length (L_{lower}) is substantially equal or longer than a length of the bore.

One advantage resides in providing a magnetic resonance imaging apparatus having a reduced bore length as seen by the patient, but with a much higher gradient efficiency than for a prior art short-bore gradient coil allowing a comparable bore shape above the patient support.

Another advantage resides in efficient use of space in the horizontally extended bottom section of a short-bore magnetic resonance imaging system.

Yet another advantage resides in providing an asymmetric short gradient coil that provides symmetric gradients in two transverse in-plane directions.

5   Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

10   The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURES 1A and 1B diagrammatically show side and end views, respectively, of a
15   magnetic resonance imaging system including a short asymmetric magnetic field gradient coil.

FIGURE 2 shows a perspective view of a wire-frame representation of the coils support of the short asymmetric magnetic field gradient coil of FIGURE 1.

FIGURE 3 shows a perspective view of the wire-frame representation of FIGURE
20   2 with the primary and shield coils layout superimposed thereon.

FIGURE 4 shows an end view of the coils support of FIGURE 2 with coil windings configured to produce two transverse gradients that are symmetrically arranged with respect to a bilateral symmetry of the magnetic field gradient coil.

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With reference to FIGURES 1A and 1B, a magnetic resonance imaging scanner **8** includes a housing **10** defining a generally cylindrical scanner bore seen by the patient. Main magnetic field generating coils **12** define a magnet bore **14** contains a scanner imaging volume **16** (indicated diagrammatically by a cross-hatched area) over which the
30   magnetic resonance imaging scanner **8** is configured to image. The housing includes an imaging subject support section **20** disposed below the imaging volume **16**. A human or other imaging subject is introduced into the scanner bore for imaging, typically on a pallet **22** or other support that slides into the scanner bore on an upper substantially planar

surface of the imaging subject support section **20** defining the bottom of the bore. The magnet bore **14** has a bore length L_{bore} along a longitudinal direction, and a diameter D_{bore} transverse to the longitudinal direction. In a preferred embodiment, the bore length L_{bore} is about the same as or shorter than the bore diameter D_{bore}. That is, for a short bore scanner

5   the L_{bore}/D_{bore} ratio is less than or about unity.

The subject support section **20** has a longitudinal length $L_{support}$ that is larger than the bore length L_{bore}. The shorter bore length L_{bore} reduces patient claustrophobia and facilitates insertion and manipulation of interventional instruments such as a catheter during imaging. The housing **10** contains or supports a radio frequency coil **28** (shown in

10   dotted phantom lines), a magnetic field gradient coil **30** (shown in dashed phantom lines), and the main magnetic field generating coils **12**, which are typically cryoshrouded superconducting coils. The magnetic field gradient coil **30** is used to generate transverse gradients, such as gradients in the x-y plane. An additional z-gradient coil (not drawn in FIGURES 1A and 1B) provides magnetic field gradients along the longitudinal or

15   z-direction.

The main magnetic field generating coils **12** generate a temporally static and substantially uniform main B_0 magnetic field generally directed in the z-direction in the Cartesian coordinates system of FIGURES 1A and 1B. The radio frequency coil **28** is selectively energized to excite magnetic resonances in the imaging volume **16**. The

20   magnetic field gradient coil **30** is selectively energized to produce one or more selected magnetic field gradients in the imaging volume **16** to spatially encode the magnetic resonances. Magnetic resonance signals are generated and spatially encoded by the gradient fields. Magnetic resonance signals are received by the radio frequency coil **28** or by another coil, and are reconstructed using a Fourier transform-based reconstruction

25   algorithm, a filtered backprojection-based reconstruction, or another suitable reconstruction algorithm. The reconstructed image is displayed, printed, stored, or otherwise utilized.

The radio frequency coil **28** is in the illustrated embodiment an asymmetric coil or coil array having a substantially flat lower portion **28₁** and a circularly arcuate upper

30   portion **28₂**. The lower and upper portions **28₁**, **28₂** of the radio frequency coil **28** comport with the asymmetry of the bore and enable the coil **28** to be disposed close to the imaging volume **16**. Suitable asymmetric radio frequency coils are described, for example, in

Overweg et al., U.S. patent no. 6,462,636, and in Leussler, international publication number WO 02/095435 A1. Alternatively, a symmetric coil, such as a cylindrical birdcage coil, can be employed. Moreover, local coils can be used in conjunction with or in place of the whole-body radio frequency coil 28. In one embodiment, for example, the whole body

5   coil 28 is used to excite magnetic resonance, and one or more local receive coils are used to receive the magnetic resonance signals.

With continuing reference to FIGURES 1A and 1B and with further reference to FIGURE 2, the mechanical coil support of the magnetic field gradient coil 30 is divided into an upper section 40 and a lower section 42. The upper section 40 has a length L_{upper}

10   along the longitudinal direction (that is, the z-direction in the case of a horizontal bore magnet using the Cartesian coordinate system of FIGURES 1A, 1B, and 2). The lower section 42 has a length L_{lower} that is larger than the longitudinal length L_{upper} of the upper section 40. The upper and lower sections 40, 42 cooperatively define a coil bore 44 therein.

The upper section 40 of the coils support has an arcuate curvature C_{upper} (indicated

15   by a curved double-headed arrow in FIGURE 2) transverse to the longitudinal direction, and substantially no curvature in the longitudinal direction, although it could flare outward toward the outer ends. The upper section 40 defines an outer surface 50 and an inner, coil bore-defining surface 52. The lower section 42 of the coil support also defines an outer surface 54 and an inner, coil bore-defining surface 56. The outer surface 54 of the lower

20   section 42 has substantially the same curvature as the arcuate curvature C_{upper} of the upper section 40, so that the outer surface 50, 54 of the gradient coil 30 has a circular, oval, or other smooth cross-section. However, the inner coil bore-defining surface 56 of the lower section 42 is more planar, e.g., conformal to a lower surface of the patient pallet 22, so that the lower surface 56 is disposed closer to the imaging volume 16 than it would be if it had

25   the arcuate curvature C_{upper}. The illustrated coil 30 has bilateral symmetry with respect to a symmetry plane 58 parallel to the y-z plane (coil symmetry plane 58 is labeled in FIGURE 1B).

Although the inner coil bore-defining surface 56 of the lower section 42 is substantially planar, it may have some curvature illustrated in FIGURES 1B and 2, which

30   however is substantially less than the arcuate curvature C_{upper} of the upper section 40. In another embodiment, the inner coil bore-defining surface 56 has the arcuate curvature C_{upper} so that the inner coil bore-defining surfaces of the coil have a circular, oval, or other

smooth cross-section. The upper and lower surfaces **50, 52** of the arcuate upper section **40** are connected by a flared connecting surface **60** that has an angle other than 90° respective to the coil bore-defining surface **52**. In the illustrated embodiment, the flared surface **60** has an angle of about 45° with respect to the coil bore-defining surface **52**. The upper and lower surfaces **54, 56** of the lower section **50** are connected by a connecting surface **62** that has an angle of 90° respective to the coil bore-defining surface **56**. The illustrated embodiment which uses the flared connecting surface **60** for the arcuate upper section **40** and facilitates flaring the scanner bore outward near ends of the bore. Such flaring can further reduce patient claustrophobia by making the scanner bore seem more "open," and can improve access. However, it is also contemplated to use orthogonal connecting surfaces for both sections, or to use flared connecting surfaces for both sections, or to use an orthogonal connecting surface for the upper section and a flared connecting surface for the lower section.

With continuing reference to FIGURES 1A, 1B, and 2, and with further reference to FIGURE 3, the magnetic field gradient coil **30** includes coil windings. Specifically, primary coil windings **70, 72** are disposed on the coil bore-defining surfaces **52, 56** of the upper and lower sections **40, 42**, respectively. Shield coil windings **74, 76** are disposed on the outer surfaces **50, 54** of the upper and lower sections **40, 42**, respectively. The primary and shield windings **70, 74** disposed on the upper section **40** are limited in longitudinal distribution to the longitudinal length L_{upper} of the upper section **40**. The primary and shield windings **72, 76** disposed on the lower section **42** are longitudinally distributed across substantially the longitudinal length L_{lower} of the lower section **42**. At least some of the individual primary windings **70, 74**, may be connected to some of the shield windings **72, 76** by field-generating conductors located on the flared connecting surface **60**.

While the coil windings **70, 72, 74, 76** are described as being disposed on the outer and inner coil bore-defining surfaces of the upper and lower sections **40, 42** of the coils support, it is to be appreciated that these windings may be encapsulated or sealed in a resin or epoxy, may be covered or sealed by a dielectric shell, or the like. In the case where more than one gradient winding, such as both vertical and horizontal transverse gradient windings, are disposed on the same surface, a dielectric shell or other insulating separator is suitably disposed between the vertical and horizontal windings to ensure electrical isolation. Moreover, it is contemplated to dispose some or all the windings inside of the

coils support. Still further, while in the illustrated embodiment both primary and shield windings are included on a single coils support, it is contemplated to omit the shield windings or to move the shield windings onto a separate coils support.

The primary coil windings **70**, **72** are electrically connected in series or in another
5   electrical configuration to define one or more transverse gradient coils that when energized impose one or more transverse gradients on the main B_0 magnetic field produced by the main magnetic field generating coils **12**. The shield windings **74** are connected with corresponding primary windings **70** across the flared connecting surface **60** by connecting conductors **80**. Similarly, the primary and shield windings **72**, **76** of the lower section **42**
10   are connected across the connecting surface **62** by connecting conductors **82**, so that energizing the primary coils **70**, **72** also energizes the shield coils **74**, **76** to produce a magnetic field gradient in the scanner imaging volume **16** with limited stray magnetic fields being produced outside of the gradient coil **30**.

The larger longitudinal length of the distribution of lower coil windings **72**, **76**,
15   which are distributed substantially coextensive with the longitudinal length L_{lower} of the lower section **42** of the coil support, versus the longitudinal length of the distribution of upper coil windings **70**, **74** which are limited to the longitudinal length L_{upper} of the upper section **40** of the coil support, has certain advantages. The return conductors of the saddle coils in the extended length section **72**, **76** are further away from the working volume of
20   the gradient coil and produce less field opposing the desired gradient field. The number of turns required for a certain gradient strength at certain operating current can thus be reduced. This reduces the magnetic field at the conductors and hence the total magnetic stored energy of the coil. This reduction of the stored energy may be a factor of 2-4 compared to a short gradient cylindrical coil of constant length. The upper portion of the
25   coil **30** is limited to the longitudinal length L_{upper} of the upper section **40** of the coils support. In the illustrated embodiment, the longitudinal length L_{upper} is substantially equal to the bore length L_{bore} of the magnet bore **14**. The gradient coil **30** thus just fits into the upper portion of the housing **10**, as shown in FIGURE 1A. The longer longitudinal length L_{lower} of the lower section **42** of the coil **30** is readily accommodated by the longitudinally
30   longer imaging subject support section **20** disposed below the imaging volume **16**, which has a length $L_{support} > L_{bore}$.

Longitudinal stretching of the magnetic gradient field generating windings is particularly advantageous for scanners in which the cross-section of the magnet bore **14** is about the same size as, or larger than, the bore length L_{bore}. In such short bore configurations, it is otherwise difficult to generate strong and fast-switching gradient fields because of the inherent inefficiency of short gradient coils. The longitudinal length L_{lower} of the lower section **42** typically extends beyond the length L_{bore} of the magnet bore **14**. In scanners with longer bores, it may be beneficial to reduce the length of the upper part of the gradient coil in order to increase the openness of the system. The gradient coil with increased lower length may also be useful for head-only scanners, where the diameter of the gradient coil is so small that the shoulders of the patient do not fit into the bore.

The magnetic field gradient coil **30** is asymmetric in the vertical or y-direction, being longitudinally shorter at top than at bottom. This vertical non-uniformity can introduce gradient non-uniformity in the vertical direction. Such gradient non-uniformity is suitably compensated by using a relatively higher windings density for coil windings **70, 74** disposed on the shorter arcuate upper section **40** of the coils support, and a relatively lower windings density for coil windings **72, 76** disposed on the longer lower section **42** of the coils support. With particular reference to FIGURE 1B, in yet another approach for improving the efficiency of the magnetic field gradient coil **30**, the windings **70, 72, 74, 76** are designed to produce a zero point **86** of the vertical (y-) magnetic field gradient (indicated by cross-hairs in FIGURE 1B) that is displaced vertically upward relative to the imaging volume **16**. That is, the gradient coil windings **70, 72, 74, 76** are designed to provide a zero z-component of the field in a horizontal plane through the zero point **86** which is above the geometric center of the scanner imaging volume **16**. By moving the zero point **86** relatively closer to the short upper section **40** and relatively farther from the long lower section **42**, the relative contribution of the upper part **40** of the gradient coil **30** and hence the contribution of this part of the coil to the total magnetic stored energy are further reduced. The more efficient lower part **42** of the coil **30** has to generate more field but the total effect of the move of the zero-field plane is an improvement in the efficiency of the coil.

In designing the coil windings **70, 72, 74, 76**, any combination of the above techniques, including using a higher density of windings at top versus bottom, and/or moving the zero point vertically upward, can be used during finite element model

optimization or other design of the coil windings **70, 72, 74, 76**. In designing the coil windings **70, 72, 74, 76**, the physical geometry and dimensions of the upper and lower sections **40, 42** of the coils support is also input into the finite element model optimization or other design optimization.

5   The magnetic field gradient coil **30** was simulated using a method described in G.N. Peeren, "Stream function approach for determining optimal surface currents," Journal of Computational Physics vol. 191, no. 1, pp. 305-321 (2003). Example simulation parameters included a length-to-diameter ratio of the inner gradient coil windings of 0.7, which is suitable for use in a short bore magnetic resonance imaging scanner having a

10   magnet bore length/diameter ratio (L_{bore}/D_{bore} in FIGURES 1A and 1B) of less than unity. The stored energy of the simulated vertical transverse gradient channel was about 2 joules, and the stored energy of the simulated horizontal transverse gradient channel was about 4 joules. These simulation results are for the coil **30** in which the lower longitudinal length L_{lower} is greater than the upper longitudinal length L_{upper}. For comparison, when the coil

15   simulation was modified by reducing L_{lower} to equal L_{upper}, and setting the internal shape of the coil to a cylinder, then the stored energy of the simulated transverse gradient channels increased to about 7 joules for both vertical and horizontal channels.

          With reference to FIGURE 4, in another embodiment, a magnetic field gradient coil **100** suitably including the coils support **40, 42** has a first gradient coils set **102, 104**

20   arranged to produce a first magnetic field gradient **106**. The coil **100** also has a second gradient coils set **112, 114** arranged to produce a second magnetic field gradient **116**. The first magnetic field gradient **106** is directed at a 45° angle respective to the symmetry plane **58** of bilaterally symmetric coils support **40, 42**. Similarly, the second magnetic field gradient **116** is also directed at a 45° angle respective to the symmetry plane **58**. The

25   magnetic field gradients **106, 116** are transverse to one another.

          The embodiment of FIGURE 4 advantageously has the two transverse magnetic field gradients **106, 116** symmetrically oriented with respect to the bilateral symmetry plane **58**. As a result, in simulations the stored energy of each of the two magnetic field gradients **106, 116** is the same, specifically about 3.3 joules for the example simulation

30   parameters including a length-to-diameter ratio of 0.7.

          Comparison with FIGURES 2 and 3 demonstrates that the coils **102** of the first gradient coils set and the coils **112** of the second coils set are disposed on inner coil

bore-defining surface **52** of the upper section **40** of the coils support. That is, these coils are part of the primary coil windings **70** for embodiments generating symmetric transverse gradients **106, 116**. On the other hand, the coils **104** of the first gradient coils set and the coils **114** of the second coils set are disposed on both inner coil bore-defining surfaces **52,**

5   **56** of the upper and lower sections **40, 42** of the coils support. That is, the coils **104, 114** are part of the primary coil windings **70, 72** for embodiments employing symmetric transverse gradients **106, 116**.

Although a horizontal bore scanner has been described herein, the magnetic field gradient coils **30, 100** are readily adapted for use in a vertical magnet scanner, by

10   re-orienting the coil so that the longitudinal direction (corresponding to the z-direction in the horizontal scanner) is oriented vertically.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description.   It is intended that the invention be

15   construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.